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Accuracy Evaluation Of microTargeting™ Platforms For Deep-Brain Stimulation Using Virtual Targets

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Abstract

Deep-brain-stimulation (DBS) surgery requires implanting stimulators at target positions with submillimetric accuracy. Traditional stereotactic frames can provide such accuracy, but a recent innovation called the microTargeting™ Platform (FHC, Inc.) replaces this large, universal frame with a single-use, miniature, custom-designed platform. Both single-target and dual-target platforms are available for unilateral and bilateral procedures, respectively. In this paper their targeting accuracies are evaluated in vitro. Our approach employs “virtual targets”, which eliminates the problem of collision of the implant with the target. We implement virtual targets by mounting fiducial markers, which are not used in platform targeting, on an artificial skull and defining targets relative to the skull via that fiducial system. The fiducial system is designed to surround the targets, thereby reducing the overall effect of fiducial localization inaccuracies on the evaluation. It also provides the geometrical transformation from image to physical space. Target selection is based on an atlas of stimulation targets from a set of 31 DBS patients. The measured targeting error is the displacement between the phantom implant and the virtual target. Our results show that the microTargeting™ Platform exhibits submillimetric in vitro accuracy with a mean of 0.42 mm and a 99.9% level of 0.90 mm.

Index Terms

Accuracy; deep brain stimulation (DBS); targeting frames; targeting platforms; virtual targets

I. Introduction¹

In 1998 FDA approval was granted for deep brain stimulation (DBS) for patients with Parkinson's disease and essential tremor. DBS is a surgical procedure that treats the tremor, rigidity, and drug-induced side effects in patients with these diseases by means of electrical stimulation. Since that year, the procedure has gained considerable recognition for the treatment of movement disorders [2], [3]. It requires that a 4-contact electrode be placed within small deep-brain target nuclei such as the subthalamic nucleus (STN), which is a structure approximately 6×4×5 mm in size [4]. Because of the small size of the targets, the placement of the electrodes traditionally requires the accuracy provided by stereotactic frames. The electrode must be placed within the target nucleus for effective stimulation [5]. If the contacts are away from the target by 3–4 mm, then the stimulation is ineffective because (a) it fails to stimulate the desired group of neurons, (b) undesired areas are stimulated resulting in unpleasant stimulation, or (c) it requires higher currents to produce the desired effect, thereby reducing the battery life of the implant. When the center of the electrode is placed within about 1 mm of the targeted neurons, these problems are avoided. Thus submillimetric accuracy is critical.

DBS targets include the Ventral intermediate nucleus (Vim), the subthalamic nucleus (STN), and the Ventrocaudalis nucleus. Targets are typically selected preoperatively in CT and/or MR images. They are not clearly visible in CT or MR images and are selected based partly on the nearby structures that are clearly visible [6]. The surgeon then adjusts the target position intraoperatively based on the electrical feedback measured in the operating room and implants the electrode with the contacts surrounding the target.

A recently developed alternative to the traditional frame is a smaller, lighter platform—the STarFix™ microTargeting™ Platform (FDA 510(K), Number: K003776, Feb 23, 2001, FHC, Inc.; Bowdoin, ME, USA). Unlike the traditional frame, which requires intraoperative adjustments for every target, each platform² is fabricated so that it is pre-aimed at the targets [7], and intraoperative adjustments are necessary only if targets are changed intraoperatively. Figures 1(A) and 1(B) show platforms for unilateral and bilateral implantations respectively. Titanium anchors (WayPoint™, FHC, Inc.) are implanted into the patient's skull and act as bases for attaching the platforms to the patient (Figures 1(C) and 1(D))—three and four anchors for the single-target and dual-target platforms respectively. An MR image volume of the head is typically obtained and a CT image volume is obtained after implanting the anchors. The two images are then registered and viewed by the surgeon. The CT image is required in order to determine the positions of the anchors, which are invisible in MR, while the MR facilitates target selection and path planning. (No MR images were necessary in this study.) A platform is then designed and custom-made based on the trajectory path relative to the anchor locations automatically by computer (see the Methods section below) [8]. During the surgery the platform is mounted on the anchors to guide the implantation through a rigid guide-tube attached to the platform.

¹The virtual-target approach and our results on dual-target platforms were presented at the 2007 IEEE International Symposium on Biomedical Imaging: From Nano to Macro in April 2007 [1].

²To avoid the repeated use of the name microTargeting™ Platform throughout this paper, we will henceforth use the term “platform”.

Figure 1(E) shows a single-target platform along with guide-tubes attached to a phantom skull. The central tube (arrow in Figure 1(F)) is aimed at the planned target position, the others being used only if a new target location is chosen intraoperatively.

The customized platform was shown in vitro to have submillimetric accuracy in reaching a target by its manufacturer for FDA testing, but they reported experimental difficulties from collisions between probe and target. A subsequent IRB-approved study to measure clinical accuracy, reported a mean error for 20 implantations of 2.8 mm [7]. This error reflected the error involved both in the surgical process and in a post-operative image-registration step. In an effort to reduce the overall error, it is important to isolate individual error contributions. To isolate the platform's contribution to the error, which arises both from error in the fabrication of the platform and error in its fixation to the head, we present here an in vitro study using skull phantoms that is based on the recently introduced concept of “virtual targets” [1]. Preliminary experiments have shown that dual-target platforms have submillimetric accuracy with 0.45 mm root-mean-square (RMS) error [1]. In the current work we report the results of experiments conducted with both single-target and dual-target platforms.

II. Methods

Our goal is to measure the distance by which a probe physically misses a target chosen within a phantom skull. The target is defined relative to a coordinate system fixed in the skull. The most direct approach would be to place a physical target in the skull and measure the distance from probe to target. Indeed this direct method was employed for the FDA testing of this platform mentioned above. However, with the direct method it is common for the probe to collide with the target, thereby preventing an accurate measurement. We have overcome this problem by replacing each physical target with a *virtual target*, which is a geometrical point specified relative to a fiducial system. The fiducial system is fixed in the skull and is not used as part of the platform's targeting system. It is important to note that there is no physical object located at the virtual target. Instead, there is a set of physical fiducials fixed in the skull that provides a means to define a coordinate system. By “fiducial” we mean specifically a point within some marker that can be accurately localized both in image space and in physical space. The fiducials are far enough from the virtual target that the probe will not collide with them. Any set of three coordinates x , y , z can specify a target relative to this fiducial system and hence relative to the skull, and any such target is a virtual target. In this section we will describe this method in detail.

In our testing we use an artificial plastic skull (Anatomical Chart Company, Skokie, IL) that is cut so that we have physical access to an interior region in which DBS targets are typically found. These represent the points at which the DBS electrodes are to be placed. To enhance the rigidity of the relatively pliable plastic, which plays the role of bone, we fill the hollow skull with a ceramic casting compound (Rescor 740, Cotronics Corp., Brooklyn, NY).

A circular ring, to which 16 fiducial markers are attached, is affixed to the inside of the skull (Figure 2). The fiducials are Titanium spheres of diameter 4.4 mm, whose centers can be

localized accurately both in CT space and in physical space. Each phantom is imaged in a CT scanner, and then virtual targets are defined in image space relative to these fiducials. The 16 fiducials on the ring are positioned such that they surround the target region [9]. As mentioned above, these fiducials are not involved in the platform targeting, but are used to provide a coordinate system fixed in the skull, relative to which virtual targets can be defined. It is important to note that, as can be seen in Figure 2, neither the fiducials nor the coordinate system defined relative to them is attached to the platform. They are attached only to the skull. Thus, any error in fixation of the platform relative to the skull will be revealed as an error in positioning of the probe relative to the virtual target, which is defined relative to the fiducial coordinate system.

In order to increase the relevance of our tests to DBS surgery, we obtain target positions by manually registering each skull CT volume to a single atlas CT volume obtained from another study [10], in which a cluster of target locations has been determined from a set of 31 DBS patients. Two target points are defined for each skull—a mean cluster position of the left STN and a mean cluster of the right STN. Random perturbations selected from a normal distribution with a standard deviation chosen to match that of the cluster are then applied to these positions to produce virtual targets in the CT image space. Entry positions, while not as critical as target positions, are also chosen so as to mimic typical entry positions for DBS patients. Each phantom CT is loaded into a planning software system (microTargeting WayPoint™ Planner, FHC, Inc.), which automatically locates each of the anchor positions, permits the user to enter the target and entry positions, and designs a platform to meet the specifications. The design parameters are then emailed to a fabrication facility (FHC, Inc.), and the platform is built and shipped back to us.

A rigid probe, shown in Figure 3(A) is used to play the role of the rigid guide-tube through which the DBS electrode implant is placed. The probe is a rigid cylindrical aluminum piece 124.75 mm in length with a sphere of diameter 9.5 mm at one end and an attachment piece at the other end shaped to mate with the platform. Once the platform arrives at the testing site, it is mounted on the anchors and the entry point(s) are marked on the skull. The platform is removed, and two burr holes are drilled in the skull, each one being at least large enough (25 to 35 mm in diameter) to allow free passage of a cylindrical aluminum probe of 12.7 mm diameter from the platform into the interior of the skull regardless of its angle (Figure 3(B)). The platform is then remounted on the anchors, and the probes are inserted into the platform through the burr holes and fixed rigidly to the platform with screws (Figure 3(C)).

To comply with typical DBS approaches, the platform is constructed such that the distance of the target from the instrument mounting surface of the platform is 120 mm, and the probe is designed so that, once it is fixed to the platform, its shaft will pass through the entry point and the center of the ball at its tip will lie at the desired target position. The center of the ball at the probe tip is the representative position of the center of the DBS implant. The error of the platform is therefore defined as the physical distance from this center to the virtual target. We call this error the “target registration error” (TRE) in keeping with standard terminology from the field of image registration [11], [12].

Each virtual target is defined relative to the skull in the CT image volume, but it is necessary to know its position relative to the skull in physical space in order to measure the distance by which the physical probe misses the target. The probe can be expected to miss the target because its position is determined by the platform, whose attachment to the skull is subject to error and whose fabrication is subject to error. The physical position of the target relative to the skull is determined by means of the fiducial markers on the ring that is attached to the skull. These markers are localized both in the image and physical space. Point-based registration [11], [12] is performed between image space and physical space to calculate the position of the target in physical space from its position in image space. At the conclusion of that calculation, the position of the target relative the skull in physical space is known.

All physical space data, namely the centers of the spherical fiducial markers and the centers of the probe tips, are acquired with a Brown and Sharpe, Chameleon coordinate measuring machine (CMM) (Wright Industries, Nashville, TN, calibration 4/11/06, certificate 4112006029735005) (Figure 2). The CMM probe makes contact with five or more points on the surface of a sphere to compute its center (using algorithms and programs residing on the CMM). The platform's measured targeting error is the distance between the center of the probe tip (measured by the CMM) and the virtual target obtained by transforming the image space target point on the basis of the positions of the fiducials in the two spaces. The measured target registration error, which we label TRE_M , is computed for each target for each platform. TRE_M is not the true error, but is the magnitude of the vector sum of three error components, which arise from independent processes. One of these is the true error that we desire to measure, while the others are corrupting errors that arise from the measurement process:

1. TRE_F —which is the error in transforming the virtual target from image space to physical space caused by combined fiducial localization error (FLE) in image and physical space
2. TLE —target localization error, which is the error in the localization of the probe tip in physical space by the CMM
3. TRE_P —true error of the platform, which is the error contributed by flaws in the localization of the anchors, the fabrication of the platform based on these anchor localizations, and the mounting of the platform onto the anchors.

Thus,

$$TRE_M = TRE_F + TLE + TRE_P. \quad (1)$$

While TRE_M is the error that we measure, it is TRE_P that we wish to report. The presence of TRE_F and TLE will result in a tendency of the magnitude TRE_M , which we report in our Results section to have a mean on the order of 0.42 mm, to be an overestimate of the true magnitude of platform error, TRE_P . To minimize that problem, we have designed our measurement technique so that the first two errors are negligible in comparison with the third. Thus, we expect to have

$$\text{TRE}_M \approx \text{TRE}_P. \quad (2)$$

In order to establish that Eq. (2) is correct, we show that TRE_F and TLE are likely to be negligible by means of statistical estimates. First we note that the RMS value of TLE has been measured for the CMM machine by its manufacturer at 0.0055 mm, which is almost two orders of magnitude below our observed TRE_M , as reported below under Results. Secondly, we can also estimate TLE as follows: All the physical measurements for a skull are performed on the same day within a short period of time (less than 90 minutes) using the same CMM with no changes in the environmental conditions. Therefore, we can safely assume that the magnitude of the physical error in localizing the centers of the larger probe tips—the TLE—is approximately the same as, if not less than, that of the physical FLE made in localizing the smaller fiducial spheres on the ring. FLE can be estimated using the following equation [11], [12]:

$$\langle \text{FLE}^2 \rangle = N \langle \text{FRE}^2 \rangle / (N - 2). \quad (3)$$

where $\langle . \rangle$ has the meaning, “expected value of”, and FRE is fiducial registration error, which is the root-mean-square (RMS) distance between corresponding fiducials after a registration. The physical locations of fiducial markers are obtained using the CMM for different platforms mounted on the same skull. Point-based registration [12] can be performed between corresponding fiducial marker locations and the FRE measured. Then using Eq. (3) FLE can be computed. Using this method for all skulls, RMS FLE in physical space was found to be 0.0071 mm, which is comparable to the manufacturer’s estimate and is likewise negligible compared to TRE_M .

RMS TRE_F error can be estimated using the equation [11], [12]:

$$\langle \text{TRE}_F^2 \rangle = \frac{\langle \text{FLE}^2 \rangle}{N} \left(1 + \frac{1}{3} \sum_{k=1}^3 \frac{d_k^2}{f_k^2} \right), \quad (4)$$

where $\langle \text{FLE}^2 \rangle$ is computed using Eq. (3) using registrations between image space to physical space, N is the number of fiducials, d_k is the distance of the virtual target from the k^{th} principal axis of the fiducial set, and f_k is the RMS distance of the fiducials themselves from their k^{th} principal axis. As given in our Results section below, the result is 0.0344 mm, which is an order of magnitude below the observed mean of TRE_M . Thus both TRE_F and TLE can be expected to be an order of magnitude or more smaller than TRE_M , indicating that Eq. (2) is indeed correct. Therefore, while our results are expected to overestimate TRE_P , the overestimation is likely to be negligible.

III. Experiment

Measurements were made on twenty platforms—eight single-target platforms and twelve dual-target platforms—mounted on four skulls. Five platforms were fabricated for each of the four skulls. The single-target platforms, which are employed to perform unilateral

procedures, require three anchors; the dual-target platforms, which are employed for bilateral procedures, require four anchors. For each of the four skulls, twelve anchors were implanted with anchor placements selected by a trained neurosurgical resident, and a CT scan was obtained on a Phillips Mx8000 IDT 16 (sixteen-slice acquisition, 120 kVp, 400mA, 750 milliseconds, slice thickness = 0.75 mm, pixel sizes = 0.422, 0.404, 0.586, and 0.490 mm for skulls 1, 2, 3, and 4, respectively). On each skull we defined two target points—left STN and right STN—and two entry points. As explained in our Methods section above, the target points were based on typical DBS targets, by randomly perturbing the mean of the cluster of targets from a patient study—one cluster for left STN and one for right STN. The random perturbation was performed for each combination of skull, target, and platform using the standard deviations of the clusters, which were [0.91, 1.99, 2.05] and [1.24, 1.77, 1.76] mm along the [Left-Right, Posterior-Anterior, Superior-Inferior] directions for left STN and right STN respectively. Then, for each skull three dual-target platforms were built for the anchors, using three sets of four anchors with no anchor in common, and two single-target platforms were built—one for the left and one for the right STN target, using two sets of three anchors. All the dual-target platforms have different left STN and right STN locations. For each skull one of the dual-target platforms was mounted on the anchors, and the two entry points were marked on the skull in the vicinity of the probe mount. Then the platform was removed and burr holes were drilled at those points to provide an entry for the probe for that platform and for the other four platforms.

For each skull, each of its five platforms, was in succession attached to the anchors on the skull and target probes—two for a dual-target platform and one for a single-target platform—were outfitted (Figure 2). The skull was then rigidly attached to a holder to avoid any movement during the CMM measurements, and the holder with the attached skull was placed on the CMM table (Figure 2). The three dual-target platforms were attached and measured first, followed by the two single-target platforms.

After the holder was placed on the CMM table, the locations of the probe tips (or probe tip for a single-target platform) were obtained, and fiducials were localized as well. For each skull, upon its first placement on the table, and only for that first placement, all sixteen fiducials were localized. These sixteen points along with sixteen corresponding image points, which were obtained by localizing these same fiducials in image space, were later used in a registration step that makes it possible to transform the virtual target points in image space into their corresponding positions in physical space. The purpose of using such a large number of fiducials is to compensate somewhat for the relative inaccuracy of image localization. Next, the holder, with the skull still rigidly attached, was removed from the CMM table. The target probes and Platform 1 were removed from the skull, and Platform 2 was fitted to a different set of anchors. The same probe(s) were then attached to Platform 2, and the holder plus skull was again placed on the CMM table.

Because the holder is unlikely to be replaced on the CMM table in the same position, it is necessary to obtain measurements to permit a re-registration with image space each time the holder is placed on the table. While the registration for Platform 2 could be performed in the same way as was done for Platform 1, i.e., by obtaining the locations of all sixteen fiducials and using them in an image-to-physical-space registration, it is not necessary. Instead, after

the holder is replaced on the table with Platform 2 attached, only three of the sixteen fiducials were localized with the CMM. The registration from image space to physical space for Platform 2 was then accomplished using both these three fiducial localizations and the sixteen localizations performed for Platform 1, as illustrated in Figure 2. The registration is accomplished by composing two registration transformations: (1) a transformation based on all sixteen fiducials from image space to the physical space that maps the skull's CT image to the skull in its position on the CMM table with Platform 1 attached and (2) a transformation from physical space to physical space based on the three chosen fiducials that maps the skull in its position on the CMM table with Platform 1 attached to its position on the table with Platform 2 attached. The use of only three fiducials for the second platform reduces the measurement time considerably while causing only a negligible loss of accuracy relative to the use of all sixteen fiducials. The loss is negligible because no image localization is involved in Step (2), and the CMM accuracy of fiducial localization is an order of magnitude better than that of image accuracy of fiducial localization. The three chosen fiducials are well spread, as can be seen in Figure 2. Platforms 3, 4, and 5 were treated similarly to Platform 2. The entire process was repeated for each skull.

IV. Results

Before making our accuracy measurements, we made estimates of the errors in our measurement technique. In particular, as explained at the beginning of our Methods section, we wished to estimate TRE_F and TLE. While the manufacturer gives a value for TLE, we made our own independent measurement of it by using the measured positions of the three fiducials that were localized on the CMM for each of the four skulls for Platforms 1 through 4 in a series of point-based registrations. As mentioned in the Methods section, by using these measurements in Eq. (3), we found by this method an estimated RMS(TLE) of 0.00708 mm over all four skulls.

Next, to estimate TRE_F we began by using the positions of the sixteen fiducials that were localized both in image space and physical space (i.e., from the CMM) for each skull's Platform 1 in a registration between image and physical space to estimate the fiducial localization error (FLE), again using Eq. (3). The resulting estimated image-to-physical RMS (FLE) over all skulls was 0.127 mm. Using this value in Eq. (4) for each of the target positions gives an estimated RMS (TRE_F) of 0.0341 mm over all four skulls and both target positions. This estimate is strictly appropriate only for Platform 1, because with Platforms 2 through 5 a composition of two registrations was employed, as explained in our Methods section. To get an estimate that is appropriate for those platforms, we first used the physical-to-physical RMS (TRE_F) of 0.00708 mm for Platforms 2 through 5 for all skulls in Eq. (4) for each of the target, which results in an estimated physical-to-physical, RMS (TRE_F) of 0.00442 mm over all skulls. To estimate the measurement error resulting from the composition of the image-to-physical transformation and the physical-to-physical transformation for Platforms 2 through 4, we then combine the RMS TREs for image-to-physical and physical-to-physical in quadrature. That combination gives a total estimated RMS (TRE_F) for Platforms 2 through 4 for all skulls of 0.0344 mm. The resulting estimates from all these measurements— $RMS(TLE) = 0.00708$ mm for all platforms, $RMS(TRE_F) =$

0.0341 mm for Platform 1, and $\text{RMS}(\text{TRE}_F) = 0.0344$ mm for Platforms 2 through 4 are used only to estimate the error of our method, not to estimate the error of the platforms.

The measured platform errors and statistics are reported in Tables 1-3. Tables 1 and 2 report the results for the single-target and dual-target platforms, respectively. As we anticipated, the estimated RMS errors of our measurement technique are an order of magnitude below $\text{RMS}(\text{TRE}_F)$. Thus, Eq. (2) is correct, showing that our method overestimates TRE_P only negligibly. We analyzed the directions of the errors and found no statistically significant directional bias at $p = 0.05$. Furthermore, we found no statistically significant difference ($p = 0.22$) between using one dual-target platform or two single-target platforms for bilateral procedures.

Table 3 reports the measured error values and statistics combining both types of platforms. Assuming a normal distribution, we find that TRE_M is less than 0.80 mm at 99% and 0.90 mm at 99.9%. Thus, the platform attains submillimetric accuracy by a considerable margin.

V. Discussion

Our results show a remarkably high in vitro targeting accuracy for this device. High accuracy is critical for DBS because of the small size of the targets and the need to hit the desired target locations. As discussed above, a placement error on the order of 2-3 mm may have a substantial effect on the success of the procedure or the duration of the batteries. Other targeting techniques can be employed for DBS, including traditional stereotactic frames and frameless systems based on bone-implanted fiducial markers in intra-operative tracking systems [13]. The most similar system is the NeXframe frameless system (Medtronic, Minneapolis, Minnesota). Our observed mean of 0.42 mm and 99.9% confidence level of 0.90 mm for this platform compare very favorably to in vitro results recently reported for the NeXframe. In a 25-site test involving 560 measurements on plastic skulls similarly to the present study, the NeXframe system exhibited a mean error of 1.25 mm and a 99.9% confidence level of 4.0 mm [14].

A high level of accuracy in probe placement is an important advantage in DBS surgery, but there are other considerations as well. The use of this platform affords a number of advantages in the surgical process relative to the traditional frame: (1) The time for the surgery, physiological mapping, and stereotactic implantation of the DBS lead is shortened relative to that required for a traditional frame because image acquisition and the target planning can be done prior to the day of the surgery, eliminating the frame application and imaging steps on the day of surgery. (2) Because of the smaller size and lighter weight of the platform, there is considerable improvement in patient tolerance and in freedom of movement of the patient, who is awake during the procedure for movement evaluation. (3) Bilateral implantations can be performed during one procedure using either two single-target platforms or one dual-target platform. (4) The platform can be rigidly affixed to the head in a repeatable fashion, if, for example, a bilateral implant needs to be staged over a week or two. An advantage relative to other frameless systems is that no tracking system is required in the operating room and no time is required for fiducial localization and registration. A disadvantage of this system is the extra procedure required approximately one week before

the surgery to implant the anchors, acquire the images, and order the platform. This system also requires the acquisition of a CT image, because the anchors are invisible in MR.

Clearly a more important measure of targeting accuracy is in vivo error, but it is difficult to measure the in vivo error for these systems, or for any targeting systems, because of the relatively large errors inherent in the in vivo evaluation itself. The fundamental difficulty is the determination of the true position of the target, which requires some means for an independent determination of target position. The error in that independent determination can be considerably larger than the targeting error of the device being evaluated, resulting in an over estimate of the error of the device. In vivo measurements using the single-target version of the platform, for example, in which the independent measurement was based on a postoperative CT being registered with a pre-operative CT, yielded a mean measurement error of 2.7 mm [7]. As reported in that work, errors arising from movement of the brain relative to the skull between the pre-op and post-op CT along with errors in the registration of the two CT images and the identification of the electrode center in the post-op image all contributed to the measured error. None of these errors affects the actual accuracy of electrode placement, but they tend to add to the measured error. The same problem plagues the in vivo evaluation of traditional frames and frameless systems, whose measured in vivo means range from 3 mm to 4 mm [7], [13], [15] for traditional frames and 3 mm for the NeXframe [13].

It is inevitable that the translation from plastic skulls in the laboratory to human skulls in the operating room will encounter some additional error. For example, the fixation error associated with homogeneous plastic may be different from that of a heterogeneous organic skull, which consists of solid cortex layers enclosing a trabecular region that is largely porous. However, because human skulls are rigid, the increase is likely to be small. The error that we measure is the error in the placement of a probe by the platform at a point relative to the skull, not relative to soft tissue, and this measured error is determined by factors that are very similar in humans and in plastic skulls. The anchors, which are made of titanium, are highly visible in CT and are considerably brighter than bone. The determination of the positioning of the platform is based only on the anchors, which are rigidly affixed to the skull, not on the shape or intensity pattern of the skull itself. Thus, differences in the imaging properties of skull and plastic will have a negligible effect, if any, on the targeting accuracy. The anchors fasten rigidly to the human skull, just as they do to a plastic one. Anchor attachment is easier for the plastic skull because of the absence of overlying scalp and muscle, which obscures the view of the anchor during implantation, but this problem can be overcome through practice and proper instrumentation. Thus, while in vitro evaluation tends to underestimate true application error, the similarity to the in vivo situation suggests that the underestimation is likely to be mild. On the other hand, in vivo evaluation tends to overestimate the error, and because of the confounding errors of the independent target determination, the overestimation is likely to be severe. Thus, it can be argued that, for placement of probes in the head, in vitro measurements are as credible a predictor of in vivo targeting accuracy as in vivo measurements, if not more so, and in view of the margin with which the platform achieved submillimetric accuracy in vitro, it seems quite safe to extrapolate our results to submillimetric targeting accuracy in the OR.

VI. Conclusion

We have employed a method based on the concept of the virtual target to evaluate the accuracy of a new stereotactic frame—the microTargeting™ Platform—which is currently being used to guide surgeries for deep-brain stimulation. The virtual target eliminates the problem of collisions between the probe, which the platform is designed to place at a target, and the target itself and thus makes the evaluation more accurate. Each virtual target is defined relative to a fiducial system that is arranged such that it surrounds the target, thereby reducing the error due to fiducial localization inaccuracies. In vitro experiments were performed with virtual targets chosen within plastic models of human skulls. The positions of the targets are based on the mean target positions of 31 DBS patients, and the planning procedure is performed similarly to the planning performed for DBS surgery. Both single-target and dual-target platforms were evaluated. Our method is likely to overestimate the platform error slightly, but our statistical estimates show that the overestimation is likely to be negligible. The mean of the observed targeting errors was 0.42 mm, and, if we assume normal error distributions, then in this testing environment fewer than one in a thousand errors would be expected to reach a millimeter. While the results apply strictly only to our in vitro testing, the similarity to the in vivo situation combined with the small size of the measured errors suggests with very high probability that for DBS surgery the microTargeting™ Platform is accurate to submillimetric level.

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Figure 1.

MicroTargeting™ Platform system. (A) Single-target platform. (B) Dual-target platform. (C) Anchors used in attaching a platform to the skull. One end of the anchor is a self-tapping screw which attaches to the skull. The other end is a hexagonal head that mates with a driver. The ruler is marked in millimeters. (D) The hexagonal portion has female threads in the top to accommodate a bolt, which passes through a leg of a platform and into the anchor to attach the platform to the anchor. (E) Single-target platform along with guide-tubes mounted on a phantom skull. (F) Guide-tubes attached to the platform. The central guide-tube (arrow) is aimed at the initially chosen target. The other tubes, each at a distance of 2 mm from the central tube are used for lateral adjustments. The cluster of 5 tubes can itself be shifted radially by 3 mm (not shown).

Figure 2.

Localization of fiducials. (A) The 16 fiducial spheres in the CT image space. (B) Physical localization of the 16 fiducials for Platform 1 for a skull using the CMM. Skull is attached to a holder to avoid movement during the CMM measurements. (C) Physical localization of the 3 fiducials (highlighted using black circles) for Platform 2 for the same skull. For Platform 1, image space is mapped on to physical space using the transformation T1 obtained by registering all the 16 fiducials on the ring in physical and image space. For Platform 2, image space is mapped on to physical space by concatenating the transformations T1 and T2. T2 is obtained by registering the 3 chosen fiducials during the measurements for Platform 2 and Platform 1.

Figure 3.

Attachment of the Probe. (A) The probe is a rigid aluminum piece 124.75 mm in length and 12.7 mm in diameter with a sphere at one end (top) of diameter 9.5 mm. At its other end is a piece that mates with a hole in the platform. (B) Top of skull showing anchors and burr holes drilled into the skull. Burr holes are of a size large enough (25 to 30 mm in diameter) to accommodate the probe regardless of its angle. (C) The platform is attached to the anchors, and then either one probe (for a single-target platform) or two probes (for a dual-target platform) are placed into the hole(s) in the platform and through the burr hole(s). Three screws fix each probe to the platform via its mating piece.

Table 1
Error Values for Single-Target Platforms (All Values in mm)

	Left STN	Right STN
Skull 1	0.39	0.81
Skull 2	0.43	0.30
Skull 3	0.41	0.38
Skull 4	0.43	0.46
All 4 skulls	0.42	0.53
All 4 skulls, both targets (N=8 targets)	RMS(TRE_M)	0.47
	Mean ± SD	0.45 ± 0.15
	Max	0.81
	Min	0.30
	99%	0.85
	99.9%	0.96

Table 2
Error Values for Dual-Target Platforms (All Values in mm)

	RMS values		Mean values	
	Left STN	Right STN	Left STN	Right STN
Skull 1	0.50	0.30	0.50	0.28
Skull 2	0.49	0.46	0.46	0.46
Skull 3	0.47	0.47	0.46	0.43
Skull 4	0.28	0.38	0.27	0.36
All 4 skulls	0.45	0.41	0.42	0.38
All 4 skulls, both targets ($N=24$ targets)	RMS(TRE_M)		0.43	
	Mean $\pm$ SD		0.40 $\pm$ 0.15	
	Max		0.69	
	Min		0.14	
	99%		0.78	
	99.9%		0.89	

Table 3
Error values for all the platforms (All values in mm)

	RMS values		Mean values	
	Left STN	Right STN	Left STN	Right STN
Skull 1	0.48	0.48	0.48	0.41
Skull 2	0.47	0.43	0.45	0.42
Skull 3	0.46	0.45	0.45	0.42
Skull 4	0.32	0.40	0.31	0.38
All 4 skulls	0.44	0.44	0.42	0.41
All 4 skulls, both targets ($N=32$ targets)	RMS(TRE_M)		0.44	
	Mean $\pm$ SD		0.42 $\pm$ 0.15	
	Max		0.81	
	Min		0.14	
	99%		0.80	
	99.9%		0.90	